MR-consistent Simultaneous Reconstruction of Attenuation and Activity for non-TOF PET/MR

Thorsten Heußer[1], Christopher M Rank[1], Thomas Beyer[2], and Marc Kachelrieß[1]

[1]Medical Physics in Radiology, German Cancer Research Center (DKFZ), Heidelberg, Germany

[2]Center for Medical Physics and Biomedical Engineering, Medical University of Vienna, Austria

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Introduction

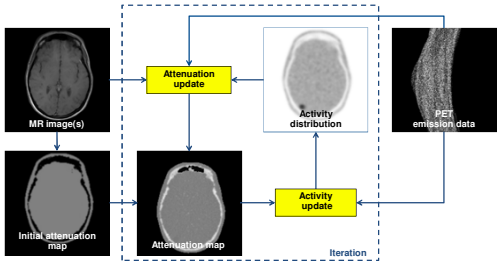
Accurate quantification of the radiotracer activity distribution in positron emission tomography (PET) mandates attenuation correction (AC). In combined PET/MR imaging, AC is a major challenge since direct conversion of the MR information into corresponding PET attenuation coefficients is not possible. The standard MR-based approach (MRAC) performs a segmentation into several tissue classes but neglects bone attenuation and, therefore, leads to an underestimation of the reconstructed PET activity distribution by up to 30%. Ongoing efforts aim at improving MRAC by considering bone attenuation derived from, e.g., ultrashort-echo-time (UTE) sequences, patient databases, or additional low-dose CT scans. In this work, we propose an extension of the maximum-likelihood reconstruction of attenuation and activity (MLAA) algorithm[1] for non-TOF PET/MR. The algorithm uses simultaneous reconstruction of attenuation and activity from the PET emission data, updating attenuation and activity in an alternating manner. Voxel-dependent expectations on the attenuation coefficients based on MR prior information are used during the attenuation update. We call the new algorithm MR-based MLAA (MR-MLAA).

Materials and Methods

MR-MLAA uses the PET emission data and available MR images as input data. The algorithm aims at simultaneously reconstructing the activity image and the attenuation image by optimizing a cost function consisting of the standard log-likelihood L and two prior terms, L_S and L_I. Solving the cost function is done by iterative optimization, alternately updating the activity while keeping the attenuation constant and vice versa. The activity update is performed by standard attenuation-weighted maximum likelihood expectation maximization (AW-MLEM) while the attenuation update is done using a gradient-descent method for transmission tomography[2]. Only the attenuation update is affected by the prior terms. The aim of the smoothing prior L_S is to favor attenuation maps which are locally smooth. It is realized by a Gibbs probability distribution penalizing differences in the attenuation values between neighboring voxels. The aim of the intensity prior is to favor the occurrence of pre-defined attenuation values. In contrast to the original MLAA, the intensity prior is realized as a local, voxel-dependent probability distribution of expected attenuation coefficients. It is defined as linear combination of the air/bone intensity prior L_{AB} and the soft tissue intensity prior L_{ST}, both of which are directly defined using piecewise linear functions as proposed in reference [1]. The voxel dependency is based on the weighting parameter $0 \le \omega(\boldsymbol{r}) \le 1$ which is given by a mask derived from the available MR images. The mask consists of two segments. One segment contains all voxels which are assumed to represent either air or bone ($\omega = 0$), the other one contains all voxels assumed to represent soft tissue ($\omega = 1$). The strength of the intensity prior relative to the likelihood function (and to the smoothing prior) can be chosen individually for the air/bone and the soft tissue segment using the parameters β_{AB} and β_{ST}, respectively. To obtain an initial attenuation map, the voxels corresponding to the air/bone segment are set to $\mu = 0$ while voxels corresponding to the soft tissue segment are set to $\mu = 0.01$ mm^{-1}. Both mask and initial attenuation map are smoothed to allow for intermediate values for ω and μ, respectively.

We evaluated our proposed algorithm for a patient data set of the head region, consisting of a T1-weighted MR image and a co-registered CT image. A 3D activity distribution including two circular 8 mm lesions with 2.5-fold activity compared to the background was simulated. 3D PET emission data were subsequently simulated accounting for Poisson noise and attenuation due to the CT image (scaled to 511 keV). For standard MR-MLAA, attenuation mask and initial attenuation map were derived from the MR images as described previously. In an idealized setting, mask and initial attenuation map were derived from the CT image. Reconstructions of the emission data were performed using the true attenuation for AC (ground truth), MRAC, and MR-MLAA, both using the standard and the idealized setting.

MR-MLAA Workflow



MR-MLAA Algorithm

Cost function:

$$C(\boldsymbol{\lambda},\boldsymbol{\mu}) = L(\boldsymbol{\lambda},\boldsymbol{\mu}) + L_S(\boldsymbol{\mu}) + L_I(\boldsymbol{\mu})$$

Standard log-likelihood | Prior terms

λ Activity; μ Attenuation; p Emission data; i Voxel index; j LOR index; n Iteration number; M System matrix; l_{ij} Intersection length of voxel i and LOR j; α Relaxation parameter

Activity update:

$$\lambda_i^{(n+1)} = \lambda_i^{(n)} \frac{1}{\sum_j M_{ij}\, a_j^{(n)}} \sum_j M_{ij} \frac{p_j}{\sum_i \lambda_i^{(n)} M_{ij}}$$

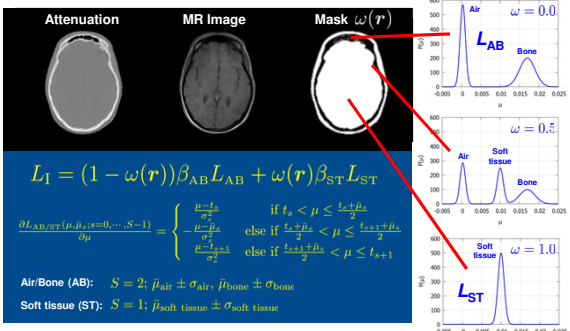
Attenuation update:

$$\mu_j^{(n+1)} = \mu_j^{(n)} + \alpha \frac{\sum_j \left(M_{ij} \left(a_j^{(n)} \sum_i \lambda_i^{(n)} M_{ij} - p_j \right) \right) + \frac{\partial}{\partial \mu_j} (L_S + L_I)}{\sum_j \left(M_{ij} a_j^{(n)} \sum_i \lambda_i^{(n)} M_{ij} \sum_i l_{ij} \right) - \frac{\partial^2}{\partial \mu_j^2} (L_S + L_I)}$$

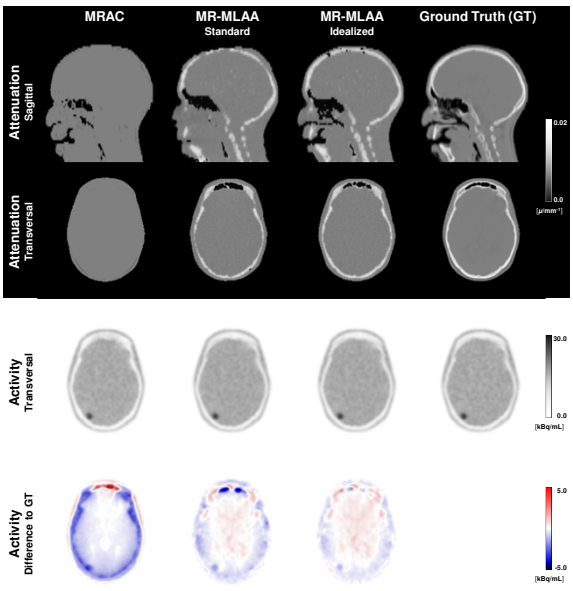
Prior

with attenuation factor $a_j^{(n)} = e^{-\sum_i \mu_i^{(n)} l_{ij}}$

Intensity Prior L_I



Results



Results

The attenuation map obtained by standard MRAC does not contain any information on bone tissue. This leads to an underestimation of the reconstructed activity, especially in the vicinity of bone. For the lesion, the average activity is underestimated by 10% compared to the ground truth. Using MR-MLAA, bone information in the attenuation map can mostly be recovered with only few misclassifications of air/bone as soft tissue and bone as air. Due to the presence of bone tissue in the MR-MLAA attenuation map, the activity estimation is greatly improved. For the lesion, the average activity is now underestimated by of 4% and 2% only, using standard and idealized MR-MLAA, respectively.

Acknowledgements

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